# Control System Design for a Prosthetic Leg Using Series Damping Actuator

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## Abstract

Bio-Mechatronics is the field that deals with passive and active prosthetic limb design. The passive conventional prosthesis carries constant mechanical properties so that the motion of joints motion is not similar to that of humans while the active type prosthesis more realistically represents human motion. The latter is however, more expensive than passive prosthesis and consumes more energy. Semi-active type prosthesis is less expensive but its results are very much similar to that of active type prosthesis and it is a better solution to control the human gait artificially. This research is based on the design of a prosthetic leg that can simulate a pattern similar to that of a normal person's gait. The research is divided into two parts: calculation of kinematics and design of control system. The forward and inverse kinematics is computed to analyze the position, orientation and workspace for the leg. Series damping actuator is used to control the swing phase of the leg and P- Controller is designed to control the swing and joint force on knee. This paper shows that a semi active prosthetic limb can emulate a real limb and in order to control a prosthetic leg Series Damping Actuator (SDA) can be used as a means to adapt the pattern of normal human gait and the walking pattern can be controlled accordingly.

**Keywords:** Kinematics, Prosthetic leg, Damping actuator, Modeling, Human gait

## Introduction

The human walking pattern is a periodic function of the movement of the upper and the lower limb. The human stance is divided into five phases namely stance flexion, stance extension, pre-swing, swing flexion and swing extension. During the stance phase the heel of the foot strikes the ground and the knee flexes and the whole weight of the body is shifted on the leg and the lower limb moves. During the swing phase the foot leaves the ground and the knee swings till the heel of the foot is ready to strike the ground [1]. The maximum energy is consumed in walk during the swing phase. The requirement of above knee (AK) prosthesis is to provide support to the knee during the stance phase and provide damping to the knee during its swing phase. The conventional AK prosthesis designs consist of different spring mass damper mechanisms but the only provide support to the leg but they did not fully anticipate in the movement of the leg [2]. The gait pattern is not similar to the normal gait as there is no active part in the mechanical design. The results of the active type prosthesis are very much similar to that of normal gait but these types of devices are expensive and consumes large amount of energy [3]. Semi Active type of prosthesis is developed that contains the properties of both active and passive type prosthesis and its result is very much similar to a normal person's gait. The semi active type devices like MR Damper consist of a mechanical part controlled by electrical part [4].

A Series Damper Actuator is a device that consists of a DC servo motor coupled with a rotary type MR Damper.MR damper contains ferromagnetic ions and in a viscous fluid and the viscosity of the fluid changes on the action of magnetic field and torque is produced. In case of SDA the MR damper acts as passive component. MR damper is used to control the inertia of the motor. The speed of the motor is controlled with the help of controller and as a result the force of the motor is controlled [5]. It is possible to generate a natural gait by controlling the knee joint by SDA.

In this research, the AK prosthesis design is modeled using SDA. Kinematics for the prosthetic leg is computed. Forward kinematics is computed to obtain the position and orientation of joints using the joint angles and inverse kinematics is done to obtain the joint angles from the orientation of the leg. Control system is designed for SDA to control the swing phase of the leg. The controller is designed for the prosthetic leg to adapt the pattern of gait of the opposite leg to obtain a natural gait.

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## Prosthetic Leg Design

### The Leg Model

The Prosthetic leg modeled as a two ling structure consisting of thigh and shank in the sagital plane. The knee joint is taken as hinge joint and the hip joint is assigned a reference ground as shown in fig 1.

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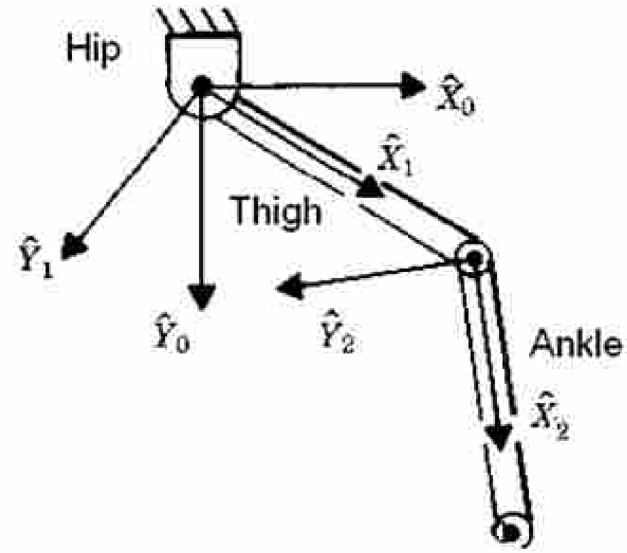


**Figure 1: The Amputees Leg Model**

$L_1$ is the length between Hip and Knee Joint
$L_2$ is the length between Knee and Ankle
$\theta_1$ is the angle between Hip and Thigh
$\theta_2$ is the angle between Thigh and Shin
$\theta_3$ is the angle between Shin and Ankle
$X_i$ is the position of the X coordinates
$Y_i$ is the position of the Y coordinates

## Forward Kinematics

The orientation and swing in the leg can be estimated using forward kinematics. The overall transformations in the leg are calculated using equation (1) [6].

$$^{i-1}T_i=\begin{bmatrix} c\theta_i & -s\theta_i & 0 & a_{i-1} \\ s\theta_i c\alpha_{i-1} & c\theta_i c\alpha_{i-1} & -s\alpha_{i-1} & -s\alpha_{i-1}d_i \\ s\theta_i s\alpha_{i-1} & c\theta_i s\alpha_{i-1} & c\alpha_{i-1} & -c\alpha_{i-1}d_i \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (1)$$

**Table 1: DH (Denavit-Heartenberg) Table**

| i | $\alpha_{i-1}$ | $a_{i-1}$ | $d_i$ | $\theta_i$ |
|---|---|---|---|---|
| 1 | 0 | 0 | 0 | $\theta_1$ |
| 2 | 0 | $L_1$ | 0 | $\theta_2$ |
| 3 | 0 | $L_2$ | 0 | $\theta_3$ |

Where
$a_i$ is the distance measured along $z_i$ and $z_{i+1}$ measured along $x_i$
$\alpha_i$ is the distance measured along $z_i$ and $z_{i+1}$ measured along $x_i$
$d_i$ is the distance measured along $x_{i-1}$ and $x_i$ measured along $z_i$
$\theta_i$ is the distance measured along $x_{i-1}$ and $x_i$ measured along $z_i$

Table 1 is substituted in equation (1) to obtain the final transformation matrix and overall workspace of the prosthetic leg given in equation (2).

$$^{0}T_3=\begin{bmatrix} A & C & 0 & E \\ B & D & 0 & F \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (2)$$

$$A = (c_1c_2 - s_1s_2)c_3 + (-c_1s_2 - s_1c_2)s_3$$
$$B = (s_1c_2 + c_1s_2)c_3 + (c_1c_2 - s_1s_2)s_3$$
$$C = -(c_1c_2 - s_1s_2)c_3 + (-c_1s_2 - s_1c_2)s_3$$
$$D = -(s_1c_2 + c_1s_2)s_3 + (c_1c_2 - s_1s_2)c_3$$
$$E = (c_1c_2 - s_1s_2)l_2 + c_1l_1$$
$$F = (s_1c_2 + c_1s_2)l_2 + s_1l_1$$

Where
$C_i = Cos\ (\theta_i)$
$S_i = Sin\ (\theta_i)$

Using equation (2) the swing of the leg is computed as given in fig 2 and the overall workspace and the swing of thigh along hip and swing of shin along knee joint is shown. Using forward kinematics the swing analysis of persons with different heights can be done and the size of the prosthetic limb can be changed according to the height of the person.

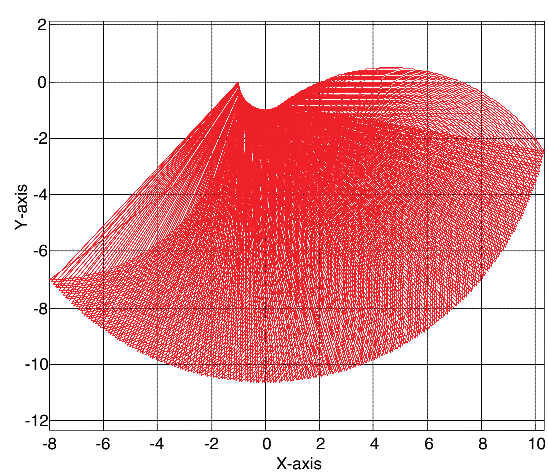


**Figure 2: Work Space of Leg**

## Inverse Kinematics

Fig 3 is the variations of joint angles obtained by using Table 1.

$$x = (L_1 + L_2c_2)c_1 - l_2s_1s_2 \quad (3)$$
$$y = (L_1 + L_2c_2)s_1 + l_2c_1s_2 \quad (4)$$
$$\varphi = \theta_1 + \theta_2 + \theta_3 = Atan2(s\varphi, c\varphi) \quad (5)$$

Using equation (3), (4) and (5) we obtain the variation of angles along the length of the leg as given in fig 3. The variations of angles in hip and knee joint are estimated.

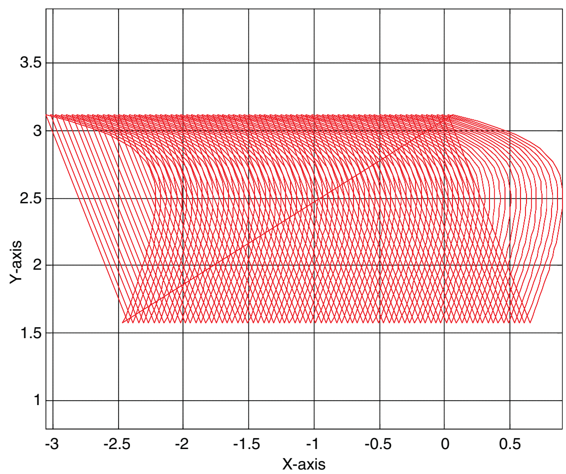


**Figure 3: Variation of Angles along Lengths**

## Control System

SDA controller only requires a unity feedback. The damper in this system is acting as passive element thus a controller to adjust the motor speed should be designed. Fig 4 shows the block diagram for the control of SDA.

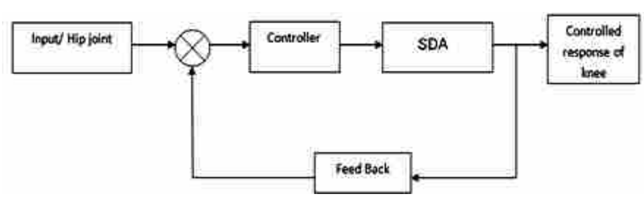


**Figure 4: Control System for SDA**

For the above controller the desired input force "$f_{desired}$" is fed to the controller and using the control system the output "$f_{load}$" is controlled at the output. Fig 5 shows the model of force control system for the prosthetic leg using SDA.

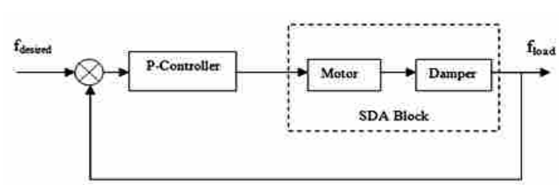


**Figure 5: P controller for SDA**

The desired input force is fed into the system and the controller computes the error in the system resulting in controlled output force and torque.

The above system is given an $F_m$ as the desired force while $F_L$ is the output force and $V_L$ is the variation in velocity in the system, $K_b$ is the damping coefficient of the series damper and $B_m$ is the damping coefficient of the motor. The sub system for SDA is given in fig 6.

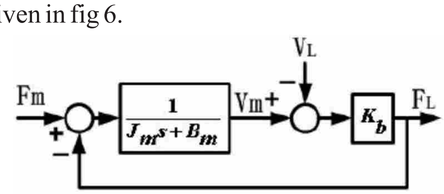


**Figure 6: SDA Sub system [5]**

The open loop response of the SDA using unit step input is given in fig 7. The response shows that open loop response of SDA is stable and settles after 2 sec.

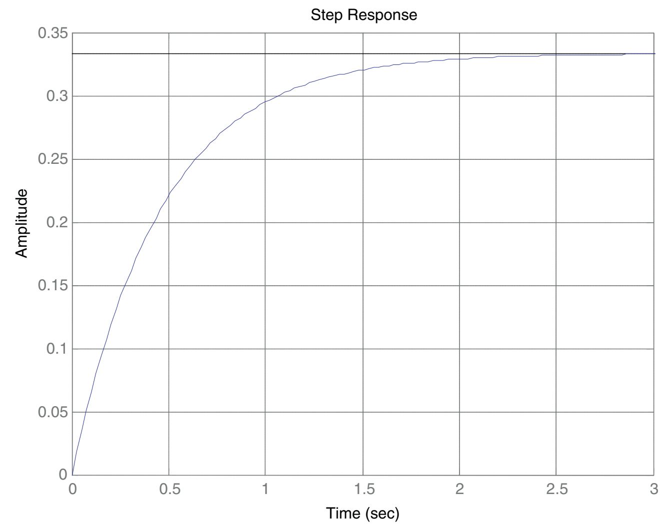


**Figure 7: Open Loop Response of SDA**

$$F_L(s) = \frac{K_p K_b F_d(s) - K_b(J_m s + B_m)V_L(s)}{J_m s + B_m + K_b(K_p + 1)} \qquad (6)$$

Equation 6 [5] is the system transfer function for the SDA block with unity feedback and P controller. "$K_p$" is the proportional controller gain.

The Root locus for the closed loop velocity input is shown in fig 8. The toot locus show that for a random velocity input the closed loop response for the velocity is stable.

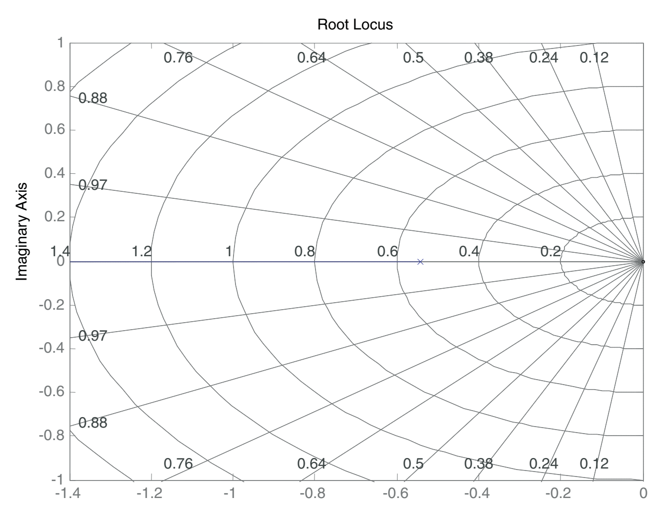


**Figure 8: Root Locus for Velocity**

The Root locus for the closed loop force input is shown in fig 9. The toot locus show that for a sinusoidal force input the closed loop response for the force on the system is stable.

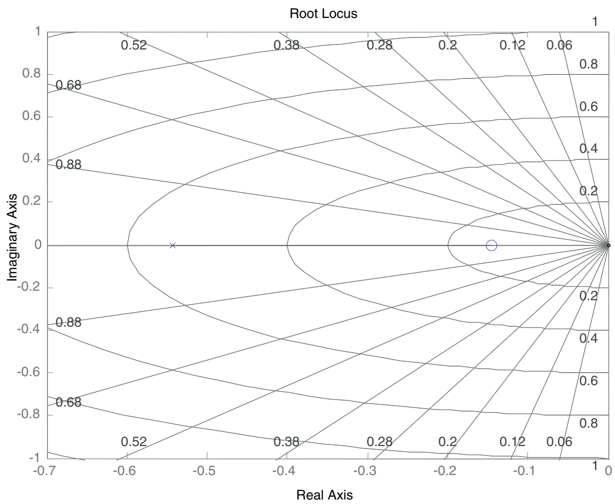


**Figure 9: Root Locus for Force**

## Gait Analysis

The system model is designed in MATLAB®. Fig 10 shows the P-Controller using SDA with velocity input to the SDA Plant. The human gait pattern is tested for various patterns of speed and force. The force is fed as an input to the P-Controller in fig 10. The speed of a person varies in walk so various patterns are used to test the working of controller.

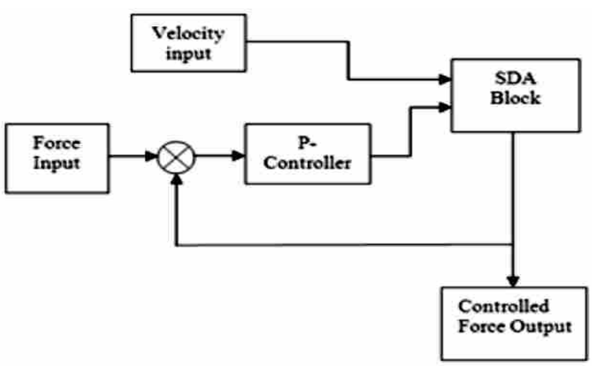


**Figure 10: SDA Controller with Velocity Input**

The SDA block consists of the transfer function due to the velocity input and the transfer function due to the force input as shown in fig 6.

First the controller is tested if a person is walking at constant speed and the force on the knee is constant so a constant unit step force input and a constant unit step velocity input is applied on the controller. As the speed of the person is constant during the gait a constant force should be applied on the leg to keep speed of the prosthetic leg constant. Ambiguities in the walking pattern occur if the speed is not kept constant resulting in a disabled person to get misbalanced.

Fig 11 shows the comparison of the input and the output force on the prosthetic leg. A constant velocity is maintained by the normal leg as shown in the fig 11 which is the result of a constant force on the leg. SDA adapts the pattern of the force on the normal leg and the output pattern of force is similar to that of the gait pattern of the normal leg.

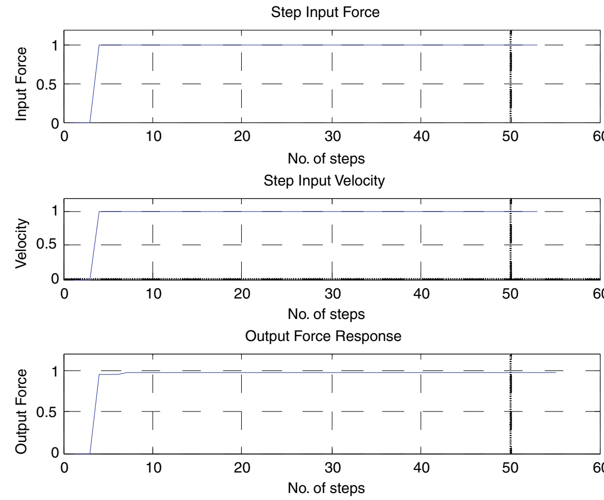


**Figure 11: Comparison of (Step) Input and Output Force**

A persons gait starts at a zero speed and increases when a person starts walking. It is required by the prosthetic leg to adapt the same pattern of the force in order to match the speed of the other leg. As the speed of the person increases, the force on the leg increases accordingly. There is no force on the leg when the person is at rest and as the gait starts the force on the knee starts increasing [7]. SDA is required to increase the speed of the motor as the persons pace starts increasing and the force generated by the SDA on the prosthetic knee should be enough to match the speed of the other leg to maintain the gait pattern. Any increase or decrease in speed results in ambiguities in the gait pattern which results in the person to lean on either leg accordingly.

Fig 12 shows the speed of the person is increasing with the number of steps taken and the force on the normal leg increases in accordance with the input speed. The controller needs the increase the speed with the increase in the input force on the normal leg. The output force in fig 12 increase by the increase in the force on the normal leg and the output force changes its pattern according the input force as the number of steps taken by the person changes. As the number of steps increases the input force increases and the output force also increases which show that the gait of both; normal and the prosthetic leg matches and the gait pattern will be stable.

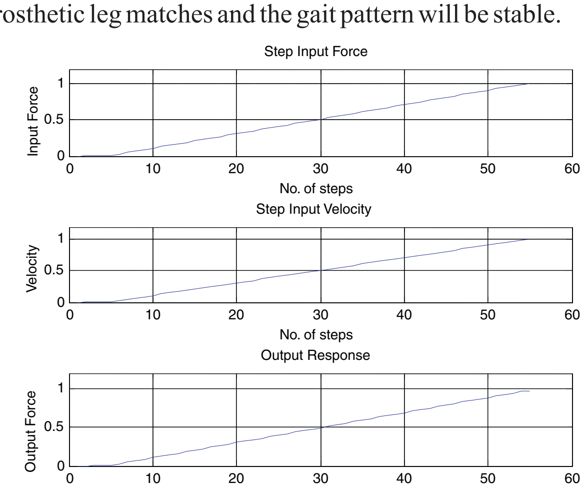


**Figure 12: Comparison of (Ramp) Input and Output Force**

The speed of person varies during walk. It increases or decreases depending upon the terrain or upon the requirement of the person. So the speed of the person can be taken as a random input on the leg. The force on the leg is a cyclic process in which one leg follows the other leg so the pattern of the input force can be taken as a sinusoidal pattern.

Fig 13 shows a random speed pattern for a person during his walk. The input force is a cyclic sinusoidal pattern in which the force on the leg increases and decreases when the leg moves forward and backwards. The speed is a random pattern and it varies on each step taken by the person.

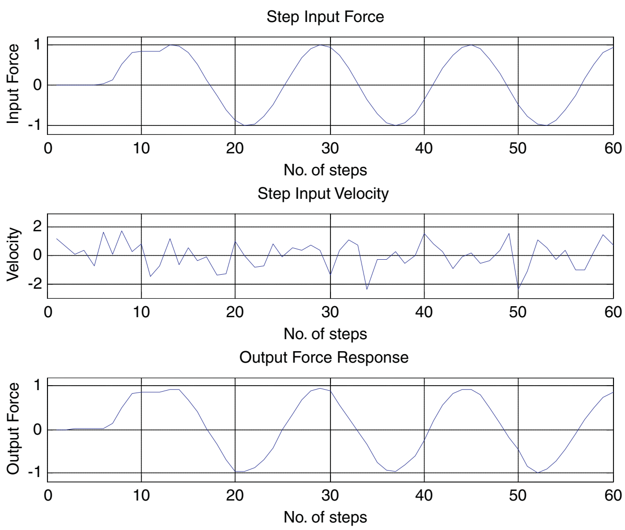


**Figure 13: Comparison of Input and Output Force Patterns**

The output force of fig 13 shows that at a random speed, the output force on the knee of the prosthetic leg adapts the same pattern as that of the input of the normal leg. Despite of a random speed the output pattern does not suffer and at any step the pattern of the output force does not deviate from the input force pattern which shows that by using SDA at a random speed the gait of person can be maintained.

## CONCLUSION

In this study, we have modeled the kinematics of the prosthetic leg. Forward and Inverse kinematics are computed by which the working space and movement of a person of any height can be computed. Forward kinematic was used to analyze the swing of the leg along its lengths and inverse kinematics was used to determine the variations in angles along the thigh and shin. We have also proposed the use of SDA as a device to act for the prosthetic leg as a substitute for knee joint in above knee AK prosthesis. We also designed the control system for SDA and analyzed the stability of the system. The controller was tested for different patterns of speed and force and it was determined that the controller adapts the pattern of gait as that of the normal leg using SDA which proves the working of controller and it is being recommended to used SDA in AK prosthesis. In comparison with MR damper the response of SDA is better as it comprises of both electrical and electromechanical systems. The response time of SDA is better than the previous work done. As for future work, SDA should be tested for different weights and the controller should be modified to adjust the torque for different weights accordingly. The error in the output pattern should be estimated and should be minimized.

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